REPORT DOCUMENTATION PAGE	*Form Approved* *OMB No. 0704-0188*

1. AGENCY USE ONLY *(Leave blank)*	2. REPORT DATE	3. REPORT TYPE AND DATES COVERED
	7.Jan.99	THESIS

4. TITLE AND SUBTITLE
DYNAMIC AND CYCLIC FATIGUE TESTING OF ESTHETIC SINGLE TOOTH IMPLANT RESTORATIONS

5. FUNDING NUMBERS

6. AUTHOR(S)
MAJ DINSE WILLIAM E

7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES)
UNIVERSITY OF TEXAS HSC SAN ANTONIO

8. PERFORMING ORGANIZATION REPORT NUMBER

9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)
THE DEPARTMENT OF THE AIR FORCE
AFIT/CIA, BLDG 125
2950 P STREET
WPAFB OH 45433

10. SPONSORING/MONITORING AGENCY REPORT NUMBER
FY99-15

11. SUPPLEMENTARY NOTES

12a. DISTRIBUTION AVAILABILITY STATEMENT
Unlimited distribution
In Accordance With AFI 35-205/AFIT Sup 1

12b. DISTRIBUTION CODE

13. ABSTRACT *(Maximum 200 words)*

19990120 019

14. SUBJECT TERMS	15. NUMBER OF PAGES
	16. PRICE CODE

17. SECURITY CLASSIFICATION OF REPORT	18. SECURITY CLASSIFICATION OF THIS PAGE	19. SECURITY CLASSIFICATION OF ABSTRACT	20. LIMITATION OF ABSTRACT

DYNAMIC AND CYCLIC FATIGUE TESTING OF ESTHETIC

SINGLE TOOTH IMPLANT RESTORATIONS

A
THESIS

Presented to the Faculty of
The University of Texas Graduate School of Biomedical Sciences
at San Antonio
in Partial Fulfillment
of the Requirements
for the Degree of

MASTER OF SCIENCE

By
William Edward Dinse, B.S., D.M.D.

San Antonio, Texas

November, 1997

DYNAMIC AND CYCLIC FATIGUE TESTING OF ESTHETIC
SINGLE TOOTH IMPLANT RESTORATIONS

William Edward Dinse

APPROVED:

Supervising Professor

24 Nov 1997
Date

APPROVED:

Sanford Miller, Ph.D.
Dean

DEDICATION

With love and admiration, I dedicate this work and the rest of my life to my wife, Joan and my children Scott, Wade, Molly, and Rebekah. Their prayers, sacrifice and encouragement helped beyond measure. I also dedicate this thesis to my father who instilled in me the desire to seek further education and my mother who gave selflessly so that her children could succeed.

ACKNOWLEDGMENTS

A heartfelt thank you goes to Dr. Stephen Schmitt for his energy and the ideas behind this project. His constant encouragement and positive attitude kept me motivated when my strength was depleted. I am grateful to Dr. Barry Norling whose fine intellect and good humor made this project almost enjoyable. Dr. Charles Hermisch unselfishly provided editorial assistance as well as help with my pilot study. Dr. David Chance kindly gave input with the initial portions of this project.

I want to acknowledge the United States Air Force Dental Corps for providing support for my graduate education in prosthodontics. This opportunity to learn and hone my clinical skills was greatly appreciated.

DYNAMIC AND CYCLIC FATIGUE TESTING OF ESTHETIC SINGLE TOOTH IMPLANT RESTORATIONS

William Edward Dinse, M.S.

The University of Texas Graduate School of Biomedical Sciences
at San Antonio

Supervising Professor: Barry K. Norling, Ph.D.

Conventional single tooth implant restorations can be unesthetic at the gingival level if they exhibit metal collars or are limited by the cylindrical shapes of the implants. Important mechanical properties of novel restorations with more esthetic features are unknown. The purpose of this investigation was to determine the elastic limit and mode of failure for four types of esthetic single tooth implant restorations and to determine the fatigue performance of these restorations. CerAdapt (CA), CeraOne (CO), and two types of non-segmented cast titanium restorations (MC and MA) were first subjected to dynamic loading to failure with an Instron machine. Four more groups were subjected to cyclic fatigue loading for 10^6 cycles. For dynamic loading, the elongation at break was significantly lower for the all ceramic restoration (CA); there were no other statistically significant differences. CA failed catastrophically in the ceramic; all others failed in the abutment screw. For cyclic loading, all four groups withstood 10^6 cycles. The low fusing ceramic used for the cast titanium restorations exhibited a high incidence of

crazing. After cyclic loading, the cast titanium restorations with margins coronal to the implant table (MC) were significantly more mobile than the other three groups. The unique cast non-segmented restorations presented in this study (MA) demonstrated comparable mechanical characteristics to the prefabricated restorations. Additional work needs to be done to improve the durability of low-fusing porcelain for titanium application. Extending the margin of a cast non-segmented restoration to the apical edge of the implant table allows more distance to create a natural emergence profile in the esthetically critical anterior region of the mouth.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

I. INTRODUCTION AND LITERATURE REVIEW

The replacement of missing anterior teeth has challenged the ingenuity of dental practitioners since the time of the Etruscans, who used ox teeth riveted to gold straps to restore missing incisors (Ring, 1986). Modern dentistry's solution to this problem for the better part of this century has been fixed and removable prosthodontics and occasionally orthodontics. The use of osseointegrated dental implants was introduced to North America in 1982 (Branemark *et al.*, 1985). Originally designed for completely edentulous patients, endosseous implants are now used to reliably replace single tooth edentulous spaces (Jemt *et al.*, 1991; Jemt and Pettersson, 1993; Cordioli *et al.*, 1994; Ekfeldt *et al.*, 1994; Anderson *et al.*, 1995). With judicious case selection and careful surgical technique, osseointegration is predictable. The emphasis has now shifted to achieving optimal esthetic results.

Single tooth implant restorations have evolved from prostheses with unsightly, metallic, barrel shaped roots to the current generation of very natural looking tooth replacements. However, the main problem associated with current anterior restorations is their tendency to be unesthetic at the gingival level. These restorations sometimes exhibit metal collars or are limited by the cylindrical geometries of the implants (Fig. 1). Human incisors have variable, individualized cervical cross sectional shapes that can range from ovoid to rectangular with rounded corners.

In select cases it is possible to fabricate implant restorations that take advantage of a greater portion of the implant by extending beyond the coronal edge of the implant table (Figs. 2 and 3). This allows a more natural, individualized emergence profile. Titanium casting and electric discharge machining (Schmitt *et al.*, 1995) can be used to fabricate these restorations.

Figure 1. Placement of a prefabricated abutment (1mm collar) in a patient with a high smile line.

A. Electric Discharge Machining

Electric discharge machining (EDM) has been used in industry for over five decades (Seman, 1975). Within the past 15 years, EDM technology has been used in dentistry for fixed and removable prosthodontics and more recently for implantology (Van Roekel, 1992; Clark, 1992; Linehan, 1994).

EDM is a method of working with metals or other electrically conductive materials to produce spots of extremely high temperature via an intermittent electric arc. This cyclic process removes minute particles of the workpiece at rates of up to 250,000 hertz. Because it is difficult to produce a micron level fit of castings using the lost wax technique, EDM was used to refine the intaglio surfaces of the cast restorations used in this project.

B. Dynamic Loading

As with many types of dental restorations such as composites, amalgams, posts and crowns, single tooth implant restorations have been subjected to dynamic loading to derive stress information for strength comparisons (Kishomoto *et al.*, 1983; Brandal *et al.*, 1987; Reeh and Douglass, 1989; Sorensen and Engelman, 1990; Trope and Tronstad, 1991). Because the single tooth implant restoration is relatively new, its dynamic loading comparison data is limited. McGlumphy's group (1992) determined that the CeramiCore (an all ceramic restoration--the forerunner of the CerAdapt) failed at 274 N when loaded in a 45 degree test fixture. Knode and Sorensen (1992) found that the CeramiCore failed at 117 N when loaded in a 30 degree test fixture. The failure data on the CeraOne (a prefabricated, titanium and ceramic restoration) ranges from 198 N (Knode and Sorensen, 1992) to 470 N (Tripodakis *et al.*, 1995). Both studies used a 30 degree test fixture. The UCLA (a non-segmented cast metal ceramic restoration)

Figure 2. Cast titanium, apically extended restoration on the same implant as Figure 1.

Figure 3. Two variations of the cast titanium restoration. The right half of the diagram shows that the margin can begin apical to the table in cases where the screw threads are exposed and more distance is required to obtain an esthetic result. The margin can originate at the coronal portion of the table (left half) similar to a UCLA restoration when the bone is at this height.

failure data ranged from approximately 198 N at 30 degrees (Knode and Sorensen, 1992) to 316 N at 45 degrees (McGlumphy *et al*., 1992).

This information is useful for designing a fatigue study. During mastication, a cyclic load is applied to restorations resulting in deformation. As long as the load does not exceed the elastic limit of the restoration, the restoration will return to its original form (Jaarda *et* al., 1995). For the purposes of this investigation, the four types of esthetic single tooth implant restorations were subjected to dynamic loading to determine their approximate elastic limits.

C. Cyclic Fatigue

Metal fatigue has been studied for many years and particularly in recent decades there has been a deluge of published work. The earliest publications appeared over 125 years ago in Britain with Wohler's classic experiments (Maydayag, 1969). These set the foundation for the fatigue testing of specimens and the determination of S/N curves. The Wohler or S-N diagram is a plot of alternating stress, S, versus cycles to failure, N. The stress-life, S-N, method was the first approach used in an attempt to understand and quantify metal fatigue. It was the standard fatigue design method for over 100 years. The S-N approach is still widely used in design applications where the resultant lives (cycles to failure) are long, such as power transmission shafts. A long life is considered over 10^5 cycles (Bannantine *et al.,* 1990).

In general, fatigue is a phenomenon that takes place in components and structures subjected to time-varying external loadings and that manifests itself in the deterioration of the material's ability to carry the intended loading. The fatigue phenomenon is thought to originate in the sliding of atomic layers of the material. This sliding is caused by a combination of local stress concentrations and dislocations. It is assumed that each slip corresponds with a small

deterioration of the material structure. Under cyclic stress conditions there is a migration of dislocations that result in localized plastic deformations. Microscopic cracks are created that grow and join together to form major cracks. Nucleation (or crack initiation) and crack growth are commonly regarded as the basic causes for ultimate fatigue failure (Sobczyk and Spencer, 1992).

The success of a dental restoration is determined in part by the ability of the material to withstand repetitive loading. It has been estimated that alternate stress fatigue applications during mastication amount to approximately 300,000 flexures per year (Craig, 1993). Evaluation of the long-term mechanical behavior of new types of dental restorations in clinical trials is time-consuming. A possible solution may be found in experimental fatigue testing, which simulates accelerated mechanical deterioration. Dental investigators have only recently begun reporting the results of various fatigue tests (Huysmans *et al.*, 1992; Gundler *et al.*, 1993; Morgan *et al.*, 1993; Wiskott *et al.*, 1994; McComb *et al.*, 1995; Basten *et al.*, 1996; Stegaroiu *et al.*, 1996).

D. Occlusal Forces

Occlusal forces on teeth and prostheses have been measured by many investigators (Anderson, 1956; Helkimo and Ingervall, 1978; Craig, 1993; Richter, 1995). Three different occlusal states can be distinguished in terms of load distribution and function of the oral system. In centric relation involving swallowing and clenching, forces are transmitted bilaterally, predominantly to molars and bicuspids. During continuous chewing, forces are usually directed unilaterally to the dentition. The third occlusal state is characterized by parafunctional motion; during grinding of the teeth, different centric or eccentric positions of the mandible with single or multiple occlusal contacts are possible. Loads applied to teeth or implants in these situations are

of abnormal quality and duration. This worst case scenario of parafunctional forces was chosen as the modality of testing for this project. The assumption being if the restorations can withstand non-axial parafunctional loading, normal chewing and swallowing forces will be easily tolerated.

During chewing, the mean load level for molars or premolars ranges between 8 and 50 N (Anderson, 1956; Graf, 1974; Lundgren, 1984). The maximum bite force measured with a bite fork between two opposing teeth is about ten times higher. Richter (1995) suggests that for chewing, only a part of the possible force development is necessary. In general, the following individual factors influence the magnitude of forces: dysgnathia (Molin, 1972), dentures in both jaws, loss of teeth and aging (Helkimino *et al.*, 1977; Lassila *et al.*, 1985). Females and dolichocephalic people develop lower muscle forces than males and brachiocephalic types (Ingervall and Helkimino, 1978; Proffitt *et al.*, 1983).

Intraoral occlusal forces applied to endosseous implants have been investigated. Haraldson *et al.* (1977), Lindquist *et al.* (1985), and Haraldson *et al.* (1988) estimated that the maximum vertical forces on mandibular Branemark endosseous implants in the premolar position ranges from 53 N to 105 N during chewing. Transducers were placed in upper dentures, actual forces to the implants were by estimation only. According to Richter (1995), the most accurate measuring modality is the use of a force transducer in the implant itself. This technique has been used in an animal model by Brunski and Hipp (1984) and more recently in humans by Setz *et al.* (1989) and Glantz *et al.* (1993). Using this technique, Richter found that implants in the molar position that were fixed to a premolar with a prosthesis withstood maximum vertical forces of 60 to 120 N during chewing. Single molars and premolars carried maximum vertical forces of 120 to 150 N.

E. Single Tooth Implant Longevity Studies

Single tooth implant restorations represent the most challenging application of the endosseous implant. There are a limited number of prospective and retrospective studies. The results of these studies are encouraging.

Jemt *et al.*, (1990) reported the three year results of single implant restorations ad modum Branemark. Sixteen patients received twenty three implants between June 1983 and May 1986. The early restorations became unstable in the screw joints and required retightening. Sixteen became loose once during the three year observation period and five became loose several times. Fistulas were observed in association with four of the loose restorations. The fistulas resolved after cleaning the components and retightening of the abutment screws. Two 7 mm implants failed on the same patient and were removed. Twenty-one implants and single crowns (91%) were in function at the end of the observation period.

In a prospective study of 40 anterior single tooth restorations Schmitt and Zarb (1993) reported a 100% success rate. The implants were loaded from 1.4 to 6.6 years. Included in their data was the first osseointegrated single-tooth replacement in North America. Occasional screw loosening was the only complication reported.

Jemt and Petterson (1993) reported on 70 single tooth implant restorations in 50 patients in a retrospective three year study. One 13 mm implant failed two months after placement of the restoration for an overall success rate of 98.5%. The long term success of the prosthetic components was not nearly as favorable. Only 13 of the 70 restorations were free of complications. Abutment screw loosening occurred at least once in 45% of the implant restorations. Fistula formation (usually associated with screw loosening) and gingival irritation

secondary to poor oral hygiene occurred less frequently. Three teeth were devitalized due to incorrect implant placement.

Cordioli *et al.*, (1994) reported follow-up clinical data on 61 single tooth implant restorations. The mean loading period was 26 months with a range of 6 to 60 months. Mechanical and soft tissue complications were reported similar to the previous studies. The overall success rate for the implants proper was 94.4%.

The CeraOne system (Nobel Biocare) was developed to address the problems associated with abutment screw instability and esthetics. Andersson (1995) reported the results of a continuing prospective study on 102 single tooth restorations using the CeraOne concept. The implant success rate was 97.3% at three years and the prosthetic success rate was 89%. The prosthetic complications were mainly traumatic fractures of the all ceramic restorations. Clearly, according to this study, the CeraOne system avoids the complication of screw loosening and associated fistula formation.

F. Investigation Objectives and Hypotheses

The objectives of this study were 1) to determine the elastic limit and mode of failure for four types of esthetic single tooth implant restorations subjected to dynamic loading and 2) to determine the fatigue performance of these restorations after subjecting them to three years of simulated non-axial parafunctional loading.

Null Hypothesis: The four single tooth implant restorations will be equivalent in terms of their elastic limits, their fracture strengths and their durability to cyclic fatigue.

Alternative Hypothesis: The four single tooth implant restorations will not be equivalent for the above three parameters.

II. MATERIALS AND METHODS

A. Experimental Plan

The purpose of this investigation was to determine the elastic limit, fracture strength and fatigue performance of four anterior single tooth restoration systems. The dynamic loading test is similar to the work of Knode and Sorensen (1992). The fatigue test is based partially on the work of Libman and Nichols (1995) and Binon (1996) although the fatigue loading device is unique to this project.

Four groups of samples were fabricated to simulate a maxillary central incisor restoration (Fig. 4). I compared the elastic limit (yield load), fracture strength (break load), and fatigue durability of an all ceramic restoration with two types of experimental cast titanium-ceramic restorations. A conventional titanium-ceramic restoration served as the control. Four groups of single tooth implant restorations were made for the dynamic load test and four groups for the fatigue test (n=6) for a total of 48 restorations.

B. Fabrication of Wax Patterns

A group of UCLA micro hex castable plastic abutment patterns (Attachments International, Inc., San Mateo, CA) was augmented with #6 green casting wax (Maves, Cleveland, OH) to an axisymetric 4.80 mm diameter and then modified to create a ferrule over the implant with wax. The margin ended at the apical aspect of the implant table (Fig. 3). This was accomplished by using a sculpting apparatus (Philp and Brukl, 1984). Another group of UCLA microhex abutment patterns was augmented with wax to an axisymetric

Figure 4. Restoration groups from left to right: Cast titanium, apically extended margin (MA); Cast titanium, margin at coronal aspect of table (MC), All alumina restoration (CA); Machined titanium restoration with cemented alumina cap--control--(CO).

4.80 mm diameter. The margin ended at the coronal aspect of the implant table (Fig 3). The internal diameter was increased to 2.60 mm using a flat ended modified drill to accommodate a 2.52 diameter CeraOne gold-palladium abutment screw.

C. Spruing and Investing

Twenty four patterns were sprued with 10 mm lengths of 6 gauge sprue wax attached to a sprue former. The patterns were sprayed with Smoothex Debubblizer (Whip Mix Corp., Louisville, KY) and air dried. The patterns were vacuum invested, two patterns to a 44.50 mm x 40.00 mm ring crucible (Whipmix), using the same batch of alumina-magnesia investment (Titavest CB, J. Morita, Japan) according to the manufacturer's instructions. The rings were next placed in a warming oven at 120°F for 120 minutes.

D. Pattern Elimination

Pattern elimination was accomplished using Titavest's recommended three phased processing cycle in a burn-out oven (Dentaurum Co., Newtown, PA). All increases and decreases in temperature were at a constant 7°C per minute. In the first phase, the temperature was brought up to 250°C and held for 20 minutes; in the second phase the temperature was raised to 830°C and held for 30 minutes; in the third phase the furnace was cooled to 600°C and held for 20 minutes prior to casting.

E. Casting

The patterns were cast in commercially pure titanium grade-1: Rematitan Ti-1 (Dentaurum Co., Newtown, PA) using a Castmatic CM330 argon pneumatic casting machine

(Iwatani Co., Osaka, Japan). Two 10 gram ingots were placed on the copper crucible in the upper chamber. The tungsten electrode was positioned 5 mm above the upper surface of the ingot. The casting ring was removed from the furnace and placed in the middle of the platform plate in the lower chamber with its opening centered below the chamber connector. The platform adjustment handle was then used to raise the ring and press it firmly into a ceramic gasket (J. Morita) placed between the ring and the top of the lower chamber around the connector.

The door was then closed, the main switch was turned on, and the START button was pressed. Argon flowed into the upper chamber to purge the room air from the system while a vacuum was being created in the lower chamber. When the argon reached 6.5 psi the arc ignited and melted the metal. After the titanium was melted and held in a molten state for 10 seconds, the support holding the crucible retracted, the crucible tipped and the melt flowed into the mold. When casting was complete, the ring was removed from the oven and quenched immediately in cool tap water. The castings were divested with an air hammer, air abraded with 50 micron aluminum oxide at 25 PSI to remove the investment and then desprued using 1.5" x 0.038" cut off discs (Jelenco, Armonk, NY).

F. Metal Finishing

The castings were finished with rubber wheel abrasives (Shofu brown and green), steam cleaned (Belle de St. Clair, Chatsworth, CA), then air abraded with 110 micron aluminum oxide at 25 PSI and steam cleaned again.

G. Electric Discharge Machining

Graphite hexed implant analogue electrodes were custom made for this project (Saturn Industries, Hudson, New York). The manufacturer matched the dimensions of a Nobel-Biocare 3.75 mm hexed implant fixture. A 10 cm piece of single strand 22 gauge copper wire was attached to each electrode. The castings were preliminarily seated on the electrodes with a gold abutment screw and the circuit was checked with an Ohm meter (model 22-218, Tandy Corp., Fort Worth, TX). The casting and electrode were attached to the ram of the Electric Discharge Machine (Hansvedt Engineering Inc., Urbana, IL). 1.0 ml of cyanoacrylate base (Zapit, DVA, Corona, CA) was dispensed with a tuberculin syringe onto the platform of the machine.

The ram was lowered until the electrode was 100 microns from the platform. 1.0 ml of cyanoacrylate accelerator was applied to the base and allowed to set for 300 seconds. The gold screw was carefully removed and the ram was raised. The EDM clips were attached to the restoration and the holding tank was filled with Rustlick EDM250 dialectric fluid (Rustlick Products, Danvers, MA). The external flushing port was placed perpendicular and 10 mm from the electrode-restoration junction. Once the electrode was beneath a volume of fluid, the casting was brought within 200 microns of electrode and the machine was switched to the autolock mode (Fig. 5). Machining proceeded until the casting seated precisely on the electrode (coronal to the table for MC and apical to the table for MA). Visual inspection with 2.5x loops were used to make this assessment.

Based on prior research (Orraca *et al.*, 1997) the machine settings were: peak current 0.195 amperes, on time 1 microsecond, flush time 5 microseconds, servo speed 4, gap adjust 2, and negative polarity. These settings provided an efficient metal removal rate with minimal electrode wear and excellent surface finish.

H. Porcelain Application

After electric discharge machining, the cast titanium abutments were steam cleaned for 30 seconds, air abraded (25 PSI, 110 micron aluminum oxide) for 10 seconds then steam cleaned for another 10 seconds and allowed to passivate 10 minutes. The titanium abutments were veneered with 1.20 mm of Duceratin low fusing porcelain made for titanium applications (Ducera Dental, Rossbach, Germany) following manufacturer's instructions for condensation sintering and firing to form a stylized central incisor (appendix). The ceramic system consists of haft-bond, gold bond, opaque, dentin, enamel and glaze. The sculpting instrument was used to obtain axisymetric, identical restorations (Fig. 6).

I. All Alumina Restorations (CA)

CerAdapt (Nobel Biocare USA, Chicago, IL) abutments were reduced to a 4.80 mm diameter with a high speed handpiece attached to the sculpting device. A medium diamond and copious water spray (15 ml/min) was used (Fig. 7). The abutments were veneered uniformly with 1.20 mm of Vitadur Alpha porcelain (Vita Zahnfabrik, Bad Sackingen, Germany) using the manufacturer's recommended firing temperatures (appendix). The sculpting instrument was used again for porcelain application and final dimension refinement.

J. Prefabricated Restorations (CO)

Six long 4.80 mm diameter ceramic caps (Nobel Biocare, DCB 128) were veneered with 1.2 mm of Vitadur Alpha porcelain using the sculpting apparatus and fired at the same schedule as CA. The caps were cemented to titanium CeraOne abutments (Nobel Biocare, DCA 121) with zinc phosphate cement (Mizzy Incorporated, Cherry Hill, New Jersey). During cementation,

Figure 5. The EDM process takes place in a dialectric fluid environment.

Figure 6 Sculpting device revolves symmetrically around the restoration resulting in uniform porcelain application.

each crown was held in place for 10 minutes under 11 pounds seating pressure, as described by Jorgensen (1960). This was accomplished by using a custom made cementation jig. Volume and viscosity of cement were controlled by standardizing the liquid-powder ratio using an Ohaus electronic E 120 scale with a readability to 0.001 grams (Ohaus Scale Corp., Florham Park, NJ). All restorations were luted during one laboratory session thereby controlling the variables of temperature and humidity.

K. Restoration Assembly

All restorations were attached to 3.75 x 20 mm implants with gold abutment screws (DCA 118, Nobel Biocare) using a controlled preload of 32 Ncm (Torque Controller, Nobel Biocare). The dynamic load group restorations were attached to a stainless steel test fixture designed to simulate the load angle of a maxillary central incisor in Angle class one occlusion (Reitz, 1972). For the fatigue test, the other group of implants were embedded in 25mm x 25mm x 33mm acrylic resin fixtures (Sampl-kwick, Buehler, Lake Bluff, IL). It has been reported that this resin approximated the resilience and elasticity of trabecular bone (Binon, 1996). The restorations were also positioned to simulate the typical 130 degree angle between the long axes of the maxillary and mandibular central incisors.

L. Dynamic Loading

Six restorations from each group were loaded at 1 mm/min. crosshead speed in air with an Instron (Fig. 8). The load-deformation data were recorded on Test Works software (Sintech, Research Triangle Park, NC). After failure occurred (fracture of the restoration) the specimens were examined at 20x under a light microscope to determine the location and mode of failure.

Figure 7. High speed handpiece attached to sculpting device for modifying CerAdapt restoration.

M. Fatigue Loading Machine

A fatigue loading machine was built by the fabrication shop at Brooks AFB specifically for this research project (Fig. 9). It cycled at 75 cycles/minute, comparable to the rate of mastication in humans (Worner, 1939; Graf, 1969; Bates *et al.*, 1975; Mohl *et al.*, 1988). At this rate, 108,000 cycles can be applied to the restoration in one 24 hour period.

The machine was a simple class 3 lever system consisting of a beam and weight that applies an intermittent load to the test specimen (Fig. 10). It had four loading stations and was driven by a Bodine 1/20 HP motor (Type NSY-34R, Bodine Electric Co., Chicago, IL). The lever arms were calibrated using the Instron machine (Appendix). The machine had a digital counter and a microswitch at each sample station to automatically stop the machine if a restoration failed.

Initially a pilot test was done with load of 50 Newtons for 1,000,000 cycles. This is a typical load applied to an anterior implant during chewing (Richter, 1995). The restorations were still intact and a load of 110 N was selected--the maximal average force generated during parafunction on an anterior implant restoration (Bo Rangert, Personal Communication, 1997). Prior to cyclic loading, all the restorations received a baseline mobility test with a Periotest machine (Siemens, West Germany). All the restorations were subjected to an intermittent 110 N force for 1 million cycles. The moment arm developed at the implant-restoration interface was 132 Ncm (110N x 1.2 cm). After the specimens were cycled, they were once again tested with the Periotest machine.

Figure 8. CerAdapt restoration prior to dynamic load testing.

Figure 9. Fatigue loading machine with four loading stations.

N. Statistical Management of Data

Data were analyzed by Analysis of Variance followed by Student-Newman-Keuls. The significance was set at $\alpha = 0.05$.

Figure 10. Alumina restoration (CA) in one of four fatigue loading stations.

III. RESULTS

A) Dynamic Loading

The data for yield, yield elongation, and break load are listed in Table 1. The yield elongation at restoration fracture was significantly lower for CA; there were no other statistically significant differences. Although there was not a statistically significant difference for break loads, the mode of fracture was different for one of the groups. The all ceramic restorations (CA) underwent brittle fracture (Fig. 11) whereas in the other three groups the implants underwent plastic deformation prior to fracture of the abutment screws (Fig. 12).

B) Cyclic Loading

All the restorations were intact after 1,000,000 cycles. A statistical summary of the Periotest values is shown in Table 2. There were no significant differences among the restoration groups prior to cyclic loading. The MC group had significantly higher Periotest values than the other three groups after cyclic loading. CA, MA, and CO were not significantly different from each other after cyclic loading. The porcelain exhibited craze lines on 58% of the cast titanium restorations (Fig. 13). The all ceramic and control restoration groups did not exhibit any porcelain flaws (Fig. 14).

Table 1. DYNAMIC LOAD MEANS AND STANDARD DEVIATIONS OF THE FOUR RESTORATION GROUPS. CA=CerAdapt, CO=CeraOne (Control), MC=CastTitanium / Margins Coronal, MA=Cast Titanium / Margins Apical, n=6.

Group	Yield, N	Yield Elongation, %	Break Load, N
CA	576 (61)	0.56 (0.04)	578 (58)
MA	581 (44)	1.08 (0.21)	594 (60)
MC	595 (75)	1.21 (0.23)	539 (38)
CO	580 (69)	1.19 (0.19)	575 (65)

Between group comparisons are by ANOVA and Student-Newman-Keuls.

Table 2. PRE AND POST CYCLIC FATIGUE PERIOTEST VALUES (PTV).

Groups n=6	Pre-fatigue PTV mean $\pm$ SD	Post-fatigue PTV mean $\pm$ SD	Increase mean $\pm$ SD	Significance of Increase
CA	6.0 ± 2.0	8.2 ± 3.2	2.2 ± 2.4	0.0401
MA	5.3 ± 2.4	10.3 ± 3.6	5.0 ± 2.0	0.0009
MC	4.4 ± 1.5	20.0 ± 9.9	15.6 ± 9.4	0.0048
CO	5.6 ± 0.7	6.9 ± 1.2	1.3 ± 1.3	0.0272

1. Significance of increase in each group is determined by the paired t test. The p-values shown are one tailed probabilities.
2. The between group comparisons are by ANOVA and Student-Newman-Keuls: no significant difference at pre-PTV stage; MC is significantly higher than the rest in post-PTV readings as well as in the increase from pre-PTV stage. The significance level used is 0.05.

Figure 11. Fractured all ceramic restoration (CA). Abutment screw is still intact.

Figure 12. Cast titanium (MC) restoration after fracture of abutment screw.

Figure 13. Cast titanium restoration exhibiting craze lines after 10^6 cycles of fatigue loading.

Figure 14. All ceramic restoration lacking evidence of craze lines after 10^6 cycles of fatigue loading.

IV. DISCUSSION AND SUMMARY

Because the single tooth implant restoration is relatively new, its dynamic loading comparison data is limited. One group determined that the CeramiCore all ceramic restoration (the forerunner of Nobel Biocare's CerAdapt restoration) failed at 274 N when loaded in a 45 degree test fixture (McGlumphy *et al.*, 1992). Others determined that the CeramiCore failed at 117 N when loaded in a 30 degree test fixture (Knode, 1992). The failure value for the present study was about double McGlumphy's and almost five times that of Knode's. One explanation for the higher values may be that the all ceramic system is second generation, with improved physical properties.

The failure data on the CeraOne ranges from 198 N according to Knode to 470 N in a study by Tripodakis *et al.*, (1995). The failure data for this project were 18% greater than Tripodakis's values.

The UCLA restoration had failure data ranging from approximately 198 N at 30 degrees (Knode) to 316 N at 45 degrees (McGlumphy). The results of this study were about 47 percent higher than McGlumphy's values. The placement of the implant in the block with one thread exposed may explain in part the increased values. Although not reported in prior studies, this placement position may have resulted in greater deflection of the restoration prior to fracture. Prior bench studies placed the implant head flush with the holding apparatus and used a variety of abutment screw types. The screw type variable was controlled in the present study by using the same gold palladium screws on all four restoration types.

The restorations in this investigation were axisymetric cylinders whereas the other three studies used tooth formed restorations. The load angle and restoration length variations between

studies no doubt resulted in lever arm variations and consequent variations in moments of torque. This would explain in part the interstudy data variability.

Although the modulus of elasticity of porcelain is moderately high, its tensile and shear strengths are low compared to titanium. As a result, only limited elastic deformation of porcelain can be tolerated. This explains the lower elongation values at break for the all ceramic restoration group. The catastrophic failure of the restoration with no observed damage to the implant may in fact provide a fail-safe mechanism for the implant itself.

When the metal ceramic restorations were subjected to dynamic non-axial loading, a recurring pattern occurred on the load/deflection curve (Fig. 15). This concurs with a recent study by Norton (1997). First the restorations showed elastic deformation, next a gap would form on the tension side of the abutment-implant interface and the implant would bend. The gap would increase as the deflection increased and ultimately the screw would fracture. The significant finding is that the screw does fracture; but not before permanent deformation of the implant itself. This finding may place some doubt on the assumption that the gold abutment screw is a fail safe mechanism.

One might expect the cast titanium restoration with the apical extension (ferrule) to exhibit significantly higher yield and fracture values than the cast titanium coronal restoration. Because the implants plastically deformed prior to screw fracture, it was difficult to draw any conclusion about this modification in the abutment geometry. Perhaps a stronger titanium alloy implant would have aided in distinguishing this difference.

Compared with dynamic loading, cyclic loading provides a closer simulation of the forces a dental restoration encounters. All the restorations endured loads equivalent to three years of simulated parafunctional tooth contact. It appears that a 110 Newton cyclic load applied to the

four groups of implant restorations is below the fatigue limit of each system. Some engineers have postulated that the fatigue limit is about one half the elastic limit of many metal alloys (Dieter, 1961; McClintock and Argon, 1966). If this held true for the test restorations, a force of about 250 Newtons would lead to fatigue fracture. More research is required to answer this question; however, forces in this range are well beyond the normal forces encountered by anterior single tooth implant restorations.

A pilot test showed that the restorations did not fracture after one million cycles therefore, Periotest values were taken before and after cyclic loading to make an objective assessment of any changes in the restoration's mobility (May *et al.,* 1997). The Periotest device resembles a handpiece with a sensor rod at its tip. When the rod is activated it taps on the restoration 16 times in 4 seconds. The sensor measures the amount of time the rod is in actual contact with the surface. The 16 measurements are measured in milliseconds and are converted to Periotest values which range from – 08 to + 50.

The reason the apically extended restorations showed less post cyclic mobility may be related to the ferrule effect. The apical margin extension moves the center of rotation apically and provides an additional vertical wall to stabilize the abutment-implant interface.

The main reason that the experimental cast titanium restorations (MC) showed Periotest values different than the control is most likely due to the loss of preload secondary to screw settling. The ability of a screw joint to obtain and maintain an adequate preload is dependent in part on fit. Debris and casting inaccuracy can negatively influence fit. One way to overcome this problem is to use wire EDM to refine the internal portion of the abutment. The application

Figure 15. Load versus deflection graph representing test specimens. Y= yield (ASTM definition: first occurrence of an increase in strain without an increase in stress resulting in a zero slope). F = load at fracture.

of this use of EDM is being examined in ongoing research. Another solution is to have a castable plastic pattern designed to accommodate the CeraOne gold screw.

The durability of the low fusing ceramic was examined. Of the twelve cast titanium restorations subjected to cyclic loading only 42 percent exhibited porcelain with no flaws. The flaws consisted mainly of vertical craze lines with some horizontal interconnecting lines. However, there was no actual loss of porcelain from the restorations. In general, the restorations from the MA group exhibited craze lines on the loading side near the base of the restoration whereas the MC group exhibited craze lines on the side opposite of the load near the base of the restoration. Extrapolating this to a clinical situation, the apically extended restoration showed craze lines on the lingual of the restoration whereas the coronal restoration exhibited craze lines on the facial of the restoration. Facial crazing would be less desirable clinically. To assess whether the craze lines originated in the ceramo-metal junction or in the porcelain itself requires further study. The apically extended group had only two defective restorations out of six whereas the coronally extended restorations had craze lines in five of the six. The reason for this difference is not clear but it may be related to the location of the center of rotation at the abutment-implant interface.

The lower modulus of elasticity of commercially pure titanium compared to ceramics may explain in part the reason for the high incidence of ceramic flaws. In addition, increased amounts of metal ions are used in low fusing ceramic to break up the silica chains. This results in a less durable ceramic. Incidentally the control restorations and the all ceramic restorations displayed no fracture or crazing. These restorations used conventional alumina which exhibits higher strength but lower translucency. Obviously, the apically extended restoration (MA) can

be cast in traditional ceramo-metal and veneered with conventional ceramic. This, in theory, should overcome the incidence of craze lines.

The results of this study differ greatly from a previous clinical study of low fusing porcelain to cast titanium crowns and fixed partial dentures (Kaus *et al.*, 1996). The authors determined a survival probability of .85 for single crowns and .59 for fixed partial dentures over a 21 to 41 month follow-up period. Survival was defined as lack of chipping or crack lines in the ceramic.

One large advantage of the cast titanium, EDM refined restorations is the potential cost savings per unit produced. Titanium is an inexpensive material and the non-segmented design obviates the need to purchase several prefabricated components. One disadvantage however, is the start up costs for the equipment.

Conclusions and Summary

Within the limits of this study the following conclusions were drawn:

1. Under dynamic loading conditions, the all ceramic restoration failed catastrophically in the ceramic; the other three restorations failed in the abutment screw.
2. For the cast titanium and machined titanium restorations the implant plastically deformed prior to ultimate failure of the abutment screw.
3. All four restoration types can withstand normal anterior biting forces but caution is advised in placing restorations of these dimensions in the posterior arch where biting forces could exceed their yield strengths.
4. All the restorations endured loads equivalent to three years of simulated parafunction.
5. Extending the restoration margin apically resulted in a lower incidence of porcelain crazing.

6. The control and the all ceramic restoration (both prefabricated) were superior to the experimental restorations in terms of post cycling mobility and incidence of minor porcelain flaws.

In summary, dynamic loading of the modified restorations exhibited similar strength characteristics as the controls while providing the advantage of recoverability and improved esthetic potential. Improved esthetics are obtained by extending the porcelain apically and creating individualized emergence profiles that simulate tooth root forms. Although the porcelain did not fracture off the restorations after three years of simulated function, the presence of craze lines calls into question the durability of the ultra-low fusing porcelain recommended for use in cast titanium restorations. More work needs to be done to enhance the durability of ultra-low fusing ceramics. This study in conjunction with future studies should aid in the development of cast titanium implant restorations that are natural in appearance, are inexpensive to manufacture, and are able to withstand the rigors of the oral environment.

APPENDIX

Dynamic Load Test: Raw Data

Restoration CA	Yield Load (N)	Elongation at Yield (mm)	Break Load (N)
1	490.328	0.522	487.973
2	617.721	0.64	613.109
3	668.469	0.576	652.708
4	572.619	0.552	567.62
5	556.951	0.535	542.95
6	552.646	0.562	602.623
mean	576.456	0.565	577.831
SD	60.909	0.041	58.071
MA			
1	497.236	1.231	522.723
2	597.112	0.818	572.301
3	574.864	0.902	573.593
4	601.232	1.118	703.284
5	586.923	1.398	586.367
6	625.641	1.003	606.727
mean	580.501	1.078	594.166
SD	44.157	0.215	60.218
MC			
1	492.641	1.175	522.016
2	524.514	0.993	493.621
3	655.775	0.983	510.112
4	667.661	1.479	591.313
5	651.506	1.485	543.391
6	576.619	1.121	572.624
mean	594.786	1.206	538.846
SD	74.765	0.226	37.599
CO			
1	533.81	0.997	532.932
2	492.027	1.011	473.463
3	690.751	1.204	662.059
4	607.018	1.503	603.066
5	599.986	1.294	600.051
6	554.623	1.115	579.604
mean	579.702	1.187	575.195
SD	69.154	0.192	64.954

Periotest Values: Raw Data

Restoration Type	Sample Number	Pre Cyclic Fatigue Values	Post Cyclic Fatigue Values
CA	1	6, 7, 8, 6, 6	6, 6, 6, 5, 5
	2	5, 6, 6, 6, 6	11, 12, 13, 13, 12
	3	4, 4, 3, 4, 5	6, 6, 6, 6, 5
	4	4, 4, 3, 4, 5	5, 5, 6, 6, 6
	5	10, 9, 10, 9, 9	12, 12, 12, 13, 12
	6	6, 6, 8, 6, 6	8,8,8,7,8
MA	1	7, 7, 8, 9, 7	16, 15, 16, 16, 16
	2	3, 3, 2, 3, 2	6, 5, 6, 6, 6
	3	7, 7, 5, 8, 7	10, 10, 10, 10, 10
	4	8, ,8, 6, 8, 9	12, 13, 14, 14, 13
	5	4, 5, 4, 4, 6	8, 8, 8, 8, 9
	6	2, 3, 3, 3, 2	9,9,9,9,9
MC	1	5, 5, 5, 5, 3	26, 26, 26, 25, 26
	2	3, 2, 3, 3, 1	22, 23, 22, 23, 23
	3	6, 6, 6, 7, 6	20, 22, 22, 23, 23
	4	5, 5, 5, 5, 5	14, 13, 16, 16, 16
	5	5, 6, 5, 6, 5	30, 32, 30, 34, 32
	6	2,3,4,4,2	3,3,3,3,3
CO	1	7, 5, 5, 5, 6	8, 8, 9, 8, 8
	2	5, ,5, 5, 6, 6	7, 7, 7, 7, 7
	3	7, 6, 6, 5, 6	8, 8, 7, 8, 8
	4	5, 4, 4, 4, 4	6, 6, 6, 6, 6
	5	5, 7, 6, 6, 6	7, 7, 8, 8,7
	6	6, 6, 6, 6, 7	5,5,5,5,5

Calibration of Cyclic Fatigue Machine (Instron, 1 KN Load Cell)

Weight (values in grams includes arm weight)	Force at Load station in newtons +/- 1N
Arm	27
125	38
250	50
375	62
500	74
625	86
750	98
1000	110
1125	122

Formula: for every 125 grams add 12 newtons of force.

Duceratin Porcelain Firing Chart

ceramic	final temp °C	predrying time (min)	heat rate (min)	holding time (min)	vacuum time (min)	Drying lift (min)
Haftbond	830	3	1	3	-	-
Goldbond	755	3	6	1	6	-
Washbake	755	3	6	1	6	-
Opaquebake	720	3	6	1	6	-
Dentinebake	710	9	6	1	6	5
Glaze	700	5	6	2	-	-

The basic temperature for all cycles was 300°C.

Vitadur Alpha Porcelain Firing Chart

Ceramic	Final Temp °C	Predrying time (min)	Heat Rate (min)	Holding Time (min)	Vacuum Time (min)
Dentin-brand	960	6	6	1	6
Korrektur-brand	950	6	6	1	6
Glanzbande	920	4	3	1	-

The basic temperature for all cycles was 600°C.

LITERATURE CITED

ANDERSON D.J. 1956a. Measurement of stress in mastication I. J. Dent. Res., 35: 664-670.

ANDERSON D.J. 1956b. Measurement of stress in mastication II. J. Dent. Res., 35: 671-673.

ANDERSSON B., ODMAN P., LINDVALL A.M., LITHNER B. 1995. Single-tooth restorations supported by osseointegrated implants: results and experiences from a prospective study after 2 to 3 years. Int. J. Oral Maxillofac. Implants, 10(6): 702- 711.

BANNATINE J.A., COMER J.J., HANDROCK J.L. 1990. Fundamentals of metal fatigue analysis. Prentice Hall, Englewood Cliffs, NJ.

BASTEN C.H., NICHOLS J.I., DALY C.H., TAGGART R. 1996. Load fatigue performance of two implant-abutment combinations. Int. J. Oral Maxillofac. Implants, 11:522-528.

BATES J.F., STAFFORD G.O., HARRISON A. 1975. Masticatory function--a review of the literature. II. Speed of movement of the mandible, rate of chewing and forces developed in chewing. J. Oral Rehab., 2:349-361.

BINON P.B. 1996. The effect of implant/abutment hexagonal misfit on screw joint stability. Int. J. Prosthodont., 9:149-160.

BRANDAL J.L., NICHOLLS J.I., HARRINGTON G.W. 1987. A comparison of three restorative techniques for endodontically treated anterior teeth. J. Prosthet. Dent., 58:161-165.

BRANEMARK P.I., ZARB G.A., ALBREKTSSON T. 1985. Tissue-integrated prosthesis: osseointegration in clinical dentistry. Quintessence Publishing Co., Chicago, IL.

BRUNSKI J.B., HIPP J.A. 1984. In vivo forces on endosteal implants: a measurement system and biomechanical considerations. J. Prosthet. Dent., 51:82.

CLARK R.S. 1992. Spark erosion technology. J. Oral. Implantol., 18:171-176.

CORDIOLI G., CASTAGNA S., CONSOLATI E. 1994. Single tooth implant rehabilitation: a retrospective study of 67 implants. Int. J. Prosthodont., 7:525-531.

CRAIG R.G. 1993. Restorative Dental Materials. 9th ed. C.V. Mosby Co., St. Louis, MO, p. 54.

DIETER G.E. 1961. Mechanical metallurgy. 1st ed. McGraw-Hill, New York, pp. 296-334.

EKFELDT A., CARLSSON G.E., BORJESSON O.H. 1994. A clinical evaluation of single-tooth restorations supported by osseointegrated implants: a retrospective study. Int.J. Oral Maxillofac. Implants, 9:179-183.

FISSORE B., NICHOLLS J.I., YOUDELIS R.A. 1991. Load fatigue of teeth restored by a dentin bonding agent and a posterior composite resin. J. Prosthet. Dent., 65:80-85.

GLANTZ P.O., RANGERT B., SVENSON A., STAFFORD G.D., ARNVIDARSON B., RNADOW K., LINDEN U., HULTEN J. 1993. On clinical loading of osseointegrated implants. A methodological and clinical study. Clin. Oral. Impl. Res., 4:99-105.

GRAF H. 1969. Bruxism. Dent. Clin. North Am., 13:659-665.

GRAF A., GRASSL H., AEBER H-J. 1974. A method for measurement of occlusal forces in three directions. Helv. Odont. Acta., 13:7-11.

GUNDLER A., LOCKOWANDT P., ERHARDSON S. 1993. Crown retention and cyclic loading (in vitro). Scand. J. Dent. Res., 101:252-256.

GUZY G.E., NICHOLLS J.I. 1979. In vitro comparison of intact endodontically treated teeth with and without endo-post reinforcement. J. Prosthet. Dent., 42:39-44.

HARALDSON T., CARLSSON G.E. 1977. Bite force and oral function in patients with osseointegrated oral implants. Scand. J. Dent. Res., 85:200-208.

HARALDSON T., ZARB G.A. 1988. 10-year follow-up study of the masticatory system after treatment with osseointegrated implant bridges. Scand. J. Dent. Res., 96:243-252.

HELKIMO E., INGERVALL B. 1978. Bite force and functional state of the masticatory system in young men. Swed. Dent. J., 2:167-175.

HUYSMANS M.C., VAN DER VARST P.G., SCHAFER R., PETERS M.C., PLASSCHAERT A.J., SOLTESZ U. 1992. Fatigue behavior of direct post-and-core-restored premolars. J. Dent. Res., 71:1145-1150.

INGERVALL B., HELKIMO E. 1978. Masticatory muscle force and facial morphology in man. Arch. Oral. Biol., 23:203-206.

JARDA M.J., RAZZOOG M.E., GRATTON D.G. 1995. Geometric comparison of five interchangeable implant prosthetic retaining screws. J. Prosthet. Dent., 74:373-379.

JEMT T., PETTERSSON P. 1993. A 3-year follow-up study on single implant treatment. J. Dent., 21:203-208.

JEMT T., LANEY W., HARRIS D. 1991. Osseointegrated implants for single tooth replacement. A one year report from a multicenter prospective study. Int. J. Oral Maxillofac. Implants, 6:29-36.

JEMT T., LEKHOLM U., GRONDAHL K. 1990. A 3-year followup study of early single implant restorations ad modum branemark. Int. J. Oral Maxillofac. Implants, 5:341-349.

JORGENSEN K.D. 1960. Factors affecting the film thickness of zinc phosphate cements. Acta. Odontol. Scand., 18:479.

KAUS T., PROBSTER L., WEBER H.P. 1996. Clinical follow-up study of ceramic veneered titanium restorations three year results. Int. J. Prosthodont., 9:9-15.

KISHIMOTO M., SHILLINGBURG H.T., DUNCANSON M.G. 1983. Influence of preparation features on retention and resistance. Part I: MOD onlays. J. Prosthet. Dent., 49:35-39.

KNODE H. 1992. Fracture strength of ceramic single tooth implant restorations. J. Dent. Res., 71: 248. (Abstract No. 1137).

LASSILA V., HOLMLUND J., KOIVUMAA K.K. 1985. Bite force and its correlations in different denture types. Acta. Odontol. Scand., 43:127-132.

LINDQUIST L.W., CARLSSON G.E. 1985. Long-term effects on chewing with mandibular fixed prostheses on osseointegrated implants. Acta. Odontol. Scand., 43:39-45.

LIBMAN W.J., NICHOLLS J.I. 1995. Load fatigue of teeth restored with cast posts and cores and complete crowns. Int. J. Prosthodont., 8:155-161.

LINEHAN A.D., WINDELER A.S. 1994. Passive fit of implant-retained prosthetic superstructures improved by electric discharge machining. J. Prosthodont. 3:88-95.

LUNDGREN D., LAURELL L. 1984. Occlusal forces in prosthetically restored dentitions: a methodological study. J. Oral. Rehabil., 11:29-37.

MADAYAG A.F. 1969. Metal fatigue: theory and design. John Wiley & Sons, Inc., New York, p. 110.

MAY K.B., EDGE M.J., RUSSELL M.M., RAZZOOG M.E., LANG B.R. 1997. The precision of fit at the implant prosthodontic interface. J. Prosthet. Dent., 77:497-502.

McCLINTOCK F.A., ARGON A.S. 1966. Mechanical behavior of materials. Addison-Wesley Publishing, Reading, Mass., pp. 576-613.

MCCOMB D.E., BROWN J.R., FORMAN F.J. 1995. Shear bond strength of resin-mediated amalgam-dentin attachment after cyclic loading. Oper. Dent., 20:236-240.

McGLUMPHY E.A., WALL J.G., ELFERS C.L., INGBER A., PRESTIPINO V. 1992. New ceramic core implant abutments; a comparison study. J. Dent. Res., 71:115. (Abstract No. 74).

MOHL N.D., ZARB G.A., CARLSSON G.E., RUGH J.D. 1988. A textbook of occlusion. Chicago: Quintessence Publ. Co., 143-152.

MOLIN C. 1972. Vertical muscle forces of the mandible. Acta. Odontol. Scand., 30:485-499.

MORGAN M.J., JAMES D.F., PILLIAR R.M. 1993. Fractures of the fixture component of an osseointegrated implant. Int. J. Oral Maxillofac. Implants, 8:409-414.

NORTON M.R. 1997. An *in vitro* evaluation of the strength of an internal conical interface compared to a butt joint interface in implant design. Clin. Oral. Impl. Res., 8:290-298.

ORRACA G.E., NORLING B.K., SCHMITT S.M., CHANCE D.A. 1997. Effect of electric discharge machining (EDM) on metal surface finish. J. Dent. Res., 76:181. (Abstract No. 1341).

PHILP G.K., BRUKL C.E. 1984. Compressive strengths of conventional, twin foil, and all-ceramic crowns. J. Prosthet. Dent., 52:215-220.

PROFFIT W.R., FIELDS H.W., NIXON W.L. 1983. Occlusal forces in normal and long-face adults. J. Dent. Res., 62:566.

REEH E.S., DOUGLAS W.H. 1989. Stiffness of endodontically treated teeth related to restoration technique. J. Dent. Res., 68:1540-1544.

REITZ P.V., AOKI H., YOSHIOKA M., UEHARA J., KUBOTA Y. 1972 . A cephalometric study of tooth position as related to facial structure in profiles of human beings: A comparison of Japanese (Oriental) and American (Caucasian) adults. J. Prosthet. Dent., 29:157-166.

RICHTER E.J. 1995. *In vivo* vertical forces on implants. Int. J. Oral. Maxillofac. Implants, 10:99-108.

RING M.E. 1986. Dentistry - An Illustrated History. C.V. Mosby Co., St. Louis, MO., pp. 43-45.

SCHMITT A., ZARB G.A. 1993. The longtitudinal clinical effectiveness of osseointegrated dental implants for single-tooth replacement. Int. J. Prosthodont., 6:197-202.

SCHMITT S.M., CHANCE D.A., CRONIN R.J. 1995. Refining cast-implant restorations by electric discharge machining. J. Prosthet. Dent., 73:280-283.

SEMAN G. 1975. A Practical Guide to Electro-Discharge Machining. Ateliers Des Charmilles SA, Geneva, pp. 4-6.

SETZ J., KRAMER A., BENSING U., WEBER H.P. 1989. Complete dentures fixed on dental implants: chewing patterns and implant stress. Int. J. Oral Maxillofac. Implants, 4:107-111.

SOBCZYK K., SPENCER B.F. 1992. Random fatigue: from data to theory. Academic Press, San Diego, CA, p. 9.

SORENSEN J.A., ENGELMAN M.J. 1990. Effect of post adaptation on fracture resistance of endodontically treated teeth. J. Prosthet. Dent., 64:417-424.

STEGAROIU R.A., YAMADA H. 1996. Retention and failure mode after cyclic loading in two post and core systems. J. Prosthet. Dent., 75:506-511.

TRIPODAKIS A., STRUB J., KAPPERT H.F., WITKOWSKI S. 1995. Strength and mode of failure of single implant all-ceramic abutment restorations under static load. Int. J. Prosthodont., 8:265-272.

TROPE M., TRONSTAD L. 1991. Resistance to fracture of endodontically treated premolars restored with glass ionomer cement or acid etch composite resin. J. Endod., 17:257-259.

VanROEKEL N. 1992. Electrical discharge machining in dentistry. Int. J. Prosthodont., 5:114-121.

WISKOTT H.W., NICHOLLS J.I., BELSER U.C. 1994. Fatigue resistance of soldered joints: a methodological study. Dent. Mat., 10:215-220.

WORNER H.K. 1939. Gnathodynamics. The measurement of biting forces with a new design of gnathodynamometer. Aust. Dent. J., 43:381-393.

VITA

William E. Dinse was born in Brentwood, Long Island, New York on 10 January 1961 to Frederick O. and Joan B. Dinse. He graduated from Athens High School, Athens, Pennsylvania in 1979. In 1983 he graduated from Houghton College, Houghton, New York, where he received a Bachelor of Science degree in Biology.

In 1987 he received the Doctor of Medical Dentistry degree from Temple University School of Dentistry, Philadelphia, Pennsylvania. At graduation he received the American Association of Oral and Maxillofacial Surgeons award and was inducted into the Omicron Kappa Upsilon National Honor Society.

In December, 1986, he accepted a commission in the United States Air Force. In July, 1988, he completed a one year General Practice Residency at David Grant Medical Center, Fairfield, California, and was assigned to Eielson AFB, Fairbanks, Alaska. Captain Dinse was assigned to RAF Lakenheath, England, in 1992. In July, 1995, he entered the combined Post-Doctoral Prosthodontic Program at Wilford Hall USAF Medical Center and the University of Texas Health Science Center at San Antonio.

Dr. Dinse was married on 2 July 1983 to Joan L. Barnes. They have four children: Scott William, Wade Samuel, Molly Grace, and Rebekah Faith.